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IMPROVED PTFE VASCULAR GRAFT AND METHOD OF MANUFACTURE**FIELD OF INVENTION:**

The present invention relates generally to a tubular implantable prosthesis such as vascular grafts and endoprostheses formed of porous polytetrafluoroethylene. More particularly, the present invention relates to a multi-layered tubular self-sealing graft or endoprosthesis formed from primarily expanded polytetrafluoroethylene.

BACKGROUND OF THE INVENTION:

It is well known to use extruded tubes of polytetrafluoroethylene (PTFE) as implantable intraluminal prostheses, particularly vascular grafts. PTFE is particularly suitable as an implantable prosthesis as it exhibits superior biocompatibility. PTFE tubes may be used as vascular grafts in the replacement or repair of a blood vessel as PTFE exhibits low thrombogenicity. In vascular applications, the grafts are manufactured from expanded polytetrafluoroethylene (ePTFE) tubes. These tubes have a microporous structure which allows natural tissue ingrowth and cell endothelization once implanted in the vascular system. This contributes to long term healing and patency of the graft.

Grafts formed of ePTFE have a fibrous state which is defined by interspaced nodes interconnected by elongated fibrils. The spaces between the node surfaces that is spanned by the fibrils is defined as the internodal distance (IND). A graft having a large IND enhances tissue ingrowth and cell endothelization as the graft is inherently more porous.

The art is replete with examples of microporous ePTFE tubes useful as vascular grafts. The porosity of an ePTFE vascular graft can be controlled by controlling the IND of the

microporous structure of the tube. An increase in the IND within a given structure results in enhanced tissue ingrowth as well as cell endothelization along the inner surface thereof.

However, such increase in the porosity of the tubular structure also results in reducing the overall radial tensile strength of the tube as well as reducing the ability for the graft to retain a suture placed therein during implantation. Also, such microporous tubular structures tend to exhibit low axial tear strength, so that a small tear or nick will tend to propagate along the length of the tube.

The art has seen attempts to increase the radial tensile and axial tear strength of microporous ePTFE tubes. These attempts seek to modify the structure of the extruded PTFE tubing during formation so that the resulting expanded tube has non-longitudinally aligned fibrils, thereby increasing both radial tensile strength as well as axial tear strength. U. S. Patent No. 4,743,480 shows one attempt to reorient the fibrils of a resultant PTFE tube by modifying the extrusion process of the PTFE tube.

Other attempts to increase the radial tensile, as well as axial tear strength of a microporous ePTFE tube include forming the tubular graft of multiple layers placed over one another. Examples of multi-layered ePTFE tubular structures useful as implantable prostheses are shown in U. S. Patent Nos. 4,816,338; 4,478,898 and 5,001,276. Other examples of multi-layered structures are shown in Japanese Patent Publication Nos. 6-343,688 and 0-022,792.

Artificial bypass grafts are often used to divert blood flow around damaged regions to restore blood flow. Vascular prostheses may also be used for creating a bypass shunt between an artery and vein. These bypass shunts are often used for multiple needle access, such as is required for hemodialysis treatments. These artificial shunts are preferable to using the body's veins, mainly because veins may either collapse along a puncture track or become aneurysmal, leaky or clotted, causing significant risk of pulmonary embolization.

While it is known to use ePTFE as a vascular prosthesis, and these vascular prostheses have been used for many years for vascular access during hemodialysis, there remain several

problems with these implantable ePTFE vascular access grafts. One major drawback in using ePTFE vascular grafts as access shunts for hemodialysis is that because of ePTFE's node-fibril structure, it is difficult to elicit natural occlusion of suture holes in the vascular prosthesis made from ePTFE tubing. As a result, blood cannot typically be withdrawn from an ePTFE vascular graft until the graft has become assimilated with fibrotic tissue. This generally takes 2 to 3 weeks after surgery. Furthermore, ePTFE's propensity for axial tears make it undesirable as a vascular access graft, as punctures, tears, and other attempts to access the blood stream may cause tears which propagate axially with the grain of the node fibril structure.

Providing a suitable vascular access graft has also been attempted in the prior art. Schanzer in U.S. Patent No. 4,619,641 describes a two-piece coaxial double lumen arteriovenous graft. The Schanzer graft consists of an outer tube positioned over an inner tube, the space between being filled with a self-sealing adhesive. The configuration of this coaxial tube greatly increases the girth of the graft, and limits the flexibility of the lumen which conducts blood flow. Herweck et al., in U.S. Patent No. 5,192,310 describes a self-sealing vascular graft of unitary construction comprising a primary lumen for blood flow, and a secondary lumen sharing a common sidewall with the primary lumen. A non-biodegradable self-sealing elastomeric material is disposed between the primary and secondary lumen.

While each of the above-referenced patents disclose self-sealing vascular grafts, none disclose a tubular access graft structure exhibiting enhanced radial tensile strength, as well as enhanced resistance to axial tear strength. Furthermore, the multi-layered ePTFE tubular structures and vascular access grafts of the prior art exhibit smaller microporous structure overall, and accordingly a reduction in ability of the graft to promote endothelization along the inner surface. Furthermore, Schanzer does not provide a self-sustaining resealable layer, but rather an elastomeric layer which "fills" the area between the two tubes.

It is therefore desirable to provide a self-sealing ePTFE graft for use in a human body which exhibits increased porosity especially at the inner surface thereof while retaining a high

degree of radial strength at the external surface thereof. The graft may preferably be used as a vascular access graft.

5 It is further desirous to produce an ePTFE vascular access graft which exhibits increased porosity at the outer surface thereof while retaining a high degree of radial tensile and suture retention strengths.

10 It is still further desirous to provide a self-sealing graft with increased resistance to axially propagating tears.

SUMMARY OF THE INVENTION:

15 It is an advantage of the present invention to provide a self-sealing ePTFE graft with increased resistance to axially propagating tears.

It is a further advantage of the present invention to provide a self-sealing ePTFE graft providing superior assimilation capabilities and resealable properties.

20 It is a further advantage of the present invention to provide a self-sealing ePTFE vascular graft exhibiting an enhanced microporous structure while retaining superior radial strength.

25 It is a still further advantage of the present invention to provide an ePTFE tubular structure having an inner portion exhibiting enhanced porosity and an outer portion exhibiting enhanced radial tensile strength, suture retention, and suture-hole elongation characteristics.

It is yet another advantage of the present invention to provide a multi-layered ePTFE tubular vascular graft having an inner layer which has a porosity sufficient to promote cell endothelization and an outer layer having a high degree of radial tensile strength.

It is an additional advantage of the present invention to provide a multi-layered ePTFE tubular vascular access graft having an outer layer whose porosity is sufficient to promote enhanced cell growth and tissue incorporation, hence more rapid healing, and an inner layer having a high degree of strength.

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In the efficient attainment of these and other advantages, the present invention provides a self-sealing ePTFE graft comprising a first expanded polytetrafluoroethylene (ePTFE) tubular structure having a first porosity, a second ePTFE tubular structure having a second porosity less than said first porosity, said second ePTFE tubular structure being disposed externally about said first ePTFE tubular structure to define a distinct porosity change between said first and second tubular structures, and a resealable polymer layer interposed between said first and second ePTFE tubular structures.

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In another embodiment, the present invention provides an ePTFE self-sealing graft, the graft formed of a first ePTFE tubular structure, a second ePTFE tubular structure disposed externally about said first ePTFE tubular structure, and further including a self-sustained resealable polymer layer interposed between the first and second ePTFE tubular structures.

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The ePTFE self-sealing graft preferably may be used as a vascular access graft. As more particularly described by way of the preferred embodiment herein, the first and second ePTFE tubular structures are formed of expanded polytetrafluoroethylene (ePTFE). Further, the second ePTFE tubular structure is adheringly supported over the first ePTFE tubular structure to form a composite tubular graft. The strength of this adhesion can be varied as desired to control the characteristics exhibited by the resultant composite structure.

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In its method aspect, the present invention provides a method of forming a self-sealing ePTFE graft. The method includes the steps of providing a first ePTFE tubular structure having a desired porosity and strength combination. A second ePTFE tubular structure is provided, also having the desired porosity and strength combination. The second ePTFE structure is disposed

over the first ePTFE so as to define a composite vascular graft.

The method of the present invention also provides for the positioning of an intermediate structure between the first and second ePTFE tubular structures. Examples of such structures include an additional ePTFE layer and fibers or thin films of PTFE or other suitable polymers. This intermediate structure also contributes to the resultant porosity and strength of the vascular graft. This intermediate structure can also preferably be a resealable polymer layer interposed between the first and second ePTFE tubular structures.

BRIEF DESCRIPTION OF THE DRAWINGS:

Figure 1 is a schematic longitudinal cross-section of a multi-layer ePTFE vascular graft of the present invention.

Figure 2 is a longitudinal cross-section of an alternate embodiment of the present invention producing a multi-layer ePTFE vascular graft.

Figure 3 is a scanning electron micrograph showing a cross-sectional view of a vascular graft produced using the present invention.

Figure 4 is a perspective showing of one of the tubular structures of the graft of Figure 1 over-wrapped with a layer of ePTFE tape.

Figure 5 is a cross-sectional showing of an alternate embodiment of the ePTFE vascular graft of the present invention.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS:

The prosthesis of the preferred embodiments of the present invention is a multi-layered

tubular structure which is particularly suited for use as a vascular access graft. The prosthesis is formed of extruded polytetrafluoroethylene (PTFE) as PTFE exhibits superior biocompatibility. In the present invention, a first ePTFE tubular structure having a first porosity is placed circumferentially interior to a second ePTFE tubular structure. Further, a resealable polymer layer is interposed as an intermediate structure between said first and second ePTFE tubular structures.

PTFE is particularly suitable for vascular applications as it exhibits low thrombogenicity. Tubes formed of extruded PTFE may be expanded to form ePTFE tubes where the ePTFE tubes have a fibrous state which is defined by elongate fibrils interconnected by spaced apart nodes. Such tubes are said to have a microporous structure, the porosity of which is determined by the distance between the surfaces of the nodes, referred to as the internodal distance (IND). Tubes having a large IND (greater than 40 microns) generally exhibit long term patency as the larger pores promote cell endothelization along the inner blood contacting surface. Tubes having lower IND (less than 40 microns) exhibit inferior healing characteristics, however they offer superior radial tensile and suture retention strengths desirable in a vascular graft. The present invention provides a composite tubular structure which promotes long term patency of the graft by providing for enhanced cell endothelization along the inner surface while exhibiting enhanced strength due to the presence of the outer layer.

Referring to Figures 1 and 2 of the drawings, composite graft 10 of the present invention is shown. Graft 10 is an elongate tubular structure formed of PTFE. Graft 10 includes a pair of coaxially disposed ePTFE tubes 12 and 14, tube 12 being the outer tube and tube 14 being the inner tube. A central lumen 15 extends through composite graft 10, defined further by the inner wall 14a of inner tube 14, which permits the passage of blood through graft 10 once the graft is properly implanted in the vascular system.

Each tube 12 and 14 may be formed in a separate extrusion process. The process for the paste extrusion of PTFE tubes is well known in the extrusion art. Once extruded, the tubes are expanded to form ePTFE tube. As will be described hereinbelow, the tubes are expanded using

differing process parameters (rates, deformation levels, temperatures, etc.) to develop the desired microporous structures. The specifically designed structure of the resulting composite tube has defined properties of strength and porosity which yield a graft 10 having long term patency and good healing characteristics as well as superior strength characteristics. It is also contemplated within the present invention to use PTFE which was extruded as sheets, expanded, and subsequently wrapped to form tubes. An ePTFE tape or ribbon helically wrapped into a tubular structure is also contemplated within the present invention.

The present invention is designed to produce grafts with substantially different node/fibril structures with respect to the internal and external portions of the graft which are adjacent to the internal and external graft surfaces. As an example, the inner tube 14 is designed to have relatively high IND while the outer tube 12 is designed to have a lower IND. Further, a distinct porosity change is clearly defined at the interface 13 between tubes 12 and 14. The inner tube 14 having a higher IND to allow enhanced cell endothelization, while the outer tube 12 having a lower IND provides superior strength to the overall composite.

An electron micrograph of such a structure produced according to the present invention is shown in Figure 3. The disparate IND's between the inner tube 14 and outer tube 12 are clearly evident, along with the step change in IND at the interface 13 between the inner tube 14 and outer tube 12. In this example, the strength of the interface 13 has been established by the processing conditions described below to fully adhere the inner tube 14 and outer tube together, hence preventing relative motion and providing enhanced strength.

Graft 10 of the present invention may be formed by expanding a thin wall inner tube 14 at a relatively high degree of elongation, on the order of approximately between 400 and 2000% elongation preferably from about between 700% and 900%. Tube 14 is expanded over a cylindrical mandrel (not shown), such as a stainless steel mandrel at a temperature of between room temperature and 645°F, preferably about 500°F. Tube 14 is preferably but not necessarily fully sintered after expansion. Sintering is typically accomplished at a temperature of between

645°F and 800°F preferably at about 660°F and for a time of between about 5 minutes to 30 minutes, preferably about 15 minutes. The combination of the ePTFE tube 14 over the mandrel is then employed as a second mandrel, over which outer tube 12 is expanded. The ID of the outer tube 12 is selected so that it may be easily but tightly disposed over the OD of inner tube 14. The composite structure 10 is then sintered at preferably similar parameters. The level of elongation of outer tube 12 is lower than that of inner tube 14, approximately between 200% and 500% elongation preferably about 400%. The expansion and sintering of outer tube 12 over the inner tube 14 serves to adheringly bond the interface 13 between the two tubes, resulting in a single composite structure 10.

In an alternate embodiment the outer tube ID may be less than the inner tube OD. In this embodiment, the outer tube is thermally or mechanically radially dilated to fit over the inner tube. The composite structure may then be sintered at about 660°F. Construction in this manner provides a snug fit of the tubes, and enhances the bonding interface between tubes, and also may augment recoil properties of an elastomeric intermediate layer.

As shown in Figure 3, the resulting composite structure has an inner surface defined by inner tube 14 which exhibits an IND of between 40 and 100 microns, spanned by a moderate number of fibrils. Such microporous structure is sufficiently large so as to promote enhanced cell-endothelization once blood flow is established through graft 10. Such cell-endothelization enhances the long term patency of the graft.

The outer structure, defined by outer tube 12, has a smaller microporous structure, with IND of about 15-35 microns and a substantial fibril density. Such outer structure results in an increase in the strength of the outer tube, and hence of the composite structure. Importantly, the outer surface defined by the outer tube 12 exhibits enhanced suture retention due to the smaller IND.

Furthermore, the resulting composite structure exhibits a sharp porosity change between the outer tube 12 and inner tube 14. This sharp porosity transition is achieved by providing an inner tube 14 having generally a given uniform porosity therealong and then providing a separate outer tube 14 having a resultant different porosity uniformly therealong. Thus a distinct porosity change is exhibited on either side of the interface 13 defined between inner tube 14 and outer tube 12.

In addition, the forming process described above results in a bonded interface between the inner tube 14 and outer tube 12. The interface exhibits sufficient interfacial strength resulting from the direct sintering of the outer tube 12 over inner tube 14 so as to assure complete bonding of the two tubes. The strength of the interface between the two tubes may be independently varied through selection of processing conditions and relative dimensions of precursor extruded tubes 12 and 14 as desired to yield a range of performance.

Referring now to Figures 4 and 5, a further embodiment of the present invention is shown. Tubular graft 20 is a composite structure similar to graft 10 described above. Graft 20 includes an outer tube 22 and an inner tube 24 formed generally in the manner described above. In order to further control the porosity and strength of the graft 20, especially at the interface between outer tube 22 and inner tube 24, an additional layer may be employed in combination with outer tube 22 and inner tube 24.

As specifically shown in Figures 4 and 5, an additional layer 26 may be employed between inner tube 24 and outer tube 22. Layer 26 may include a helical wrap of ePTFE tape 27 placed over inner tube 24. The additional layer 26, however, may also exist as a sheet, film, yarn, monofilament or multi filament wrap, or additional tube. The additional layer 26 may consist of PTFE, FEP, or other suitable polymer composition to obtain the desired performance characteristics. Layer 26 may be used to impart enhanced properties of porosity and/or strength to the composite graft 20. For example, an additional layer 26 of ePTFE tape 27 having a low IND and wrapped orthogonally to the length direction of graft 20 would increase the radial

strength of the resultant composite graft. Similarly, a layer of ePTFE having a high IND would increase the porosity of the composite structure thereby further promoting cell endothelization and/or tissue ingrowth.

5 In a preferred embodiment of the present invention, the intermediate layer may be a resealable polymer layer, employed in order to create a self-sealing graft. The self-sealing graft may be preferably used as a vascular access device. The graft is also preferably implantable. When used as an access device, the graft allows repeated access to the blood stream through punctures, which close after removal of the penetrating member (such as, e.g., a hypodermic
10 needle or cannula) which provided the access.

The intermediate, or resealable polymer layer may be additionally augmented with a pre-sintered PTFE (or FEP) bead wrap, or wire support coil. The pre-sintered PTFE bead wrap is a cylindrically extruded solid tube of PTFE that is sintered, then helically wrapped around the
15 desired layer (inner, intermediate, or outer). A graft of this embodiment shows enhanced strength and handling characteristics, i.e., crush resistance, kink resistance, etc.

In another preferred embodiment, a self-sustained resealable polymer layer may be interposed between first and second ePTFE tubular structures. For the purpose of this
20 specification, the term self-sustained refers to a tubular structure which possesses enough structural stability to be formed and subsequently stand alone without the use of additional tubular layers, or any other "molding" type formation, i.e., not a resinous polymer which is injected, or fills a space between an outer and inner tube. Some examples include the elastomeric layers employed in the present invention, as well as the resealable intermediate layers shown in Examples
25 3 and 4 of the present invention.

The ePTFE self-sealing graft can be used for any medical technique in which repeated hemoaccess is required, for example, but without intending to limit the possible applications, intravenous drug administration, chronic insulin injections, chemotherapy, frequent blood samples,

connection to artificial lungs, and hyperalimentation. The self-sealing ePTFE graft is ideally suited for use in chronic hemodialysis access, e.g., in a looped forearm graft fistula, straight forearm graft fistula, an axillary graft fistula, or any other AV fistula application. The self-sealing capabilities of the graft are preferred to provide a graft with greater suture retention, and also to prevent excessive bleeding from a graft after puncture (whether in venous access or otherwise).

The graft is made self-sealing with the use of a resealable polymer layer interposed between said first and second polymer layer. The resealable layer functions by primarily two different mechanisms. In one embodiment, the resealable polymer layer comprises an elastomeric component. The term elastomeric as used herein refers to a substance which is capable of essentially rebounding to near its initial form or state after deformation. In another embodiment, the resealable polymer layer comprises a flowable material layer. The term flowable as used herein refers to an amorphous material which fills a void created by a deformation or puncture.

It is further contemplated within the present invention to provide a composite vascular graft with an intermediate resealable layer, and multiple interior or exterior layers of ePTFE. Furthermore, the use of multiple intermediate layers possessing resealable properties is also contemplated within the present invention.

A number of different materials may be employed to provide an elastomeric polymer component as contemplated in the present invention. Furthermore, the elastomeric properties of the intermediate layer may be imparted thereto as a result of an inherent property of the material used, or as a result of the particular method of constructing such a layer. The elastomeric component may also be adhered to the first and second ePTFE tubular structures. The adhesion may take place by mechanical means, chemical means (use of an adhesive), either, or both. Some polymers, particularly thermoplastic elastomers, become sufficiently tacky through heating to adhere to the ePTFE tubular structures. The elastomeric component may also exert a force in the direction of the puncture, which if adhered to the first and/or second ePTFE tubular structures may provide for either layer to seal the puncture site. Some materials which may be used as an

elastomeric component in various forms include, but are not limited to, polymers and copolymers, including thermoplastic elastomers and certain silicones, silicone rubbers, synthetic rubbers, polyurethanes, polyethers, polyesters, polyamides and various fluoropolymers, including, but not limited to PTFE, ePTFE, FEP (fluorinated ethylene propylene copolymer), and PFA

5 (polyfluorinated alkanate). The materials may be utilized as the elastomeric polymer layer in a number of different forms which would impart the desired elastomeric characteristics to the layer.

In one embodiment, an extruded polymeric ribbon or tape wrap may be wrapped helically into a tubular shape under tension. Alternatively, a sheet, fiber, thread, or yarn may also be wrapped under tension to impart an elastomeric layer.

10 In another preferred embodiment, a polymeric layer may be applied from solution. The polymer may be dissolved or partially dissolved in a solvent, and upon evaporation of the solvent, the polymer is deposited as an elastomeric layer. The solvents used in this system must be capable of wetting the ePTFE tubular surfaces. Upon evaporation of the solvent, an elastomeric layer is deposited which may penetrate into the pores of the adjacent ePTFE layer to provide an anchoring effect for the polymeric layer. Upon evaporation of the solvent, the elastomeric layer may also shrink to provide the desired elastomeric characteristics.

15 In another embodiment of the elastomeric layer of the present invention, a solvent spun polyurethane as disclosed in U.S. Patent Nos. 4,810,749; 4,738,740; 4,743,252 and 5,229,431, herein incorporated by reference, may be employed. From such elastomeric fibers may be formed a woven or non-woven textile-like layer with sufficient fiber density to form a sealing layer while allowing a puncturing member such as a hypodermic needle or cannula to penetrate between the individual fibers. Furthermore, the elastomeric fibers of said textile-like structure may be 20 employed under tension or compression to facilitate the recovery of the fibers displaced by the penetrating member to their original position after removal by the penetrating member.

25 Furthermore the elastomeric layer of the present invention may additionally be impregnated with a gel to provide enhanced sealing capabilities. Examples of such gels are

hydrogels formed from natural materials including, but not limited to, gelatin, collagen, albumin, casein, algin, carboxy methyl cellulose, carageenan, furcellaran, agarose, guar, locust bean gum, gum arabic, hydroxyethyl cellulose, hydroxypropyl cellulose, methyl cellulose, hydroxyalkylmethyl cellulose, pectin, partially deacetylated chitosan, starch and starch derivatives, including amylose and amylopectin, xanthan, polylysine, hyaluronic acid, and its derivatives, heparin, their salts, and mixtures thereof.

A number of different flowable polymer layers may be interposed between said first and second tubular structures to provide a self-sealing graft. The flowable polymer layer seals the graft by possessing an amorphous quality which fills in any space left open subsequent to puncture of the graft. It may simply fill in the space left open in the interposed middle layer, or it may additionally penetrate into the first and/or second ePTFE tubular structures to fill any void left from puncture of either layer.

An example of a flowable polymer which may be used in the present invention is an uncured or partially cured polymer. The polymer may be cured by a number of activating means which would activate curing subsequent to puncture of the graft, thereby sealing with the curing of the polymer. Examples of materials for such a flowable layer include, but are not limited to, uncured elastomers such as natural or synthetic rubbers, and natural gums such as gum arabic. Materials that are particularly useful in a flowable layer include non-crosslinked polyisobutylene which is also known as uncured butyl rubber.

Another flowable polymer layer which may be employed in the present invention is a gel. Gels are generally suspensions or emulsions of polymers which have properties intermediate the liquid and solid states. A hydrogel may also be used in the present invention, and refers to polymeric material which swells in water without dissolving, and which retains a significant amount of water in its structure. The gels and hydrogels employed in the present invention may be biodegradable, or non-biodegradable. They also further may have polymeric beads (not to be confused with the pre-sintered PTFE bead-wrap, which imparts structural stability) suspended

within the gel to effectuate sealing of the prosthesis. Some examples of gels which may be used in the present invention include, but are not limited to, silicone gels, gum arabic, and low molecular weight ethylene/vinyl acetate polymers.

5 The following examples serve to provide further appreciation of the invention but are not meant in any way to restrict the scope of the invention.

EXAMPLE I

10 A thin extruded tube having wall thickness of 0.41 mm and an inner diameter of 6.2 mm was expanded over a stainless steel mandrel at 500°F to 900% elongation. The ePTFE tube was then sintered at 660°F for 14 minutes, cooled, and removed from the oven. A second thin extruded tube having wall thickness of 0.45 mm and an inner diameter of 6.9 mm was expanded over the first tube/mandrel combination at 500°F and 400% elongation. The composite was then
15 sintered at 660°F for 14 minutes, cooled and removed from the oven. The resultant composite tube had a wall thickness of 0.65 mm and ID of 5.8 mm.

EXAMPLE 2

20 A thin extruded tube having wall thickness of 0.41 mm and an inner diameter of 6.2 mm was expanded over a stainless steel mandrel at 500°F to 700% elongation. The ePTFE tube was then sintered at 660°F for 14 minutes, cooled, and removed from the oven. A second thin extruded tube having wall thickness of 0.45 mm and an inner diameter of 6.9 mm was expanded over the first tube at 500°F and 400% elongation. The composite was sintered at 660°F for 14
25 minutes, cooled, and removed from the oven. The resultant composite tube had a wall thickness of 0.67 mm and an inner diameter of 5.8 mm.

Table I presents physical property data for a vascular graft of the type depicted in Example I described above. The composite graft was removed from the mandrel and subjected to standard

testing of radial tensile strength and suture hole elongation. The radial strength of the 900%/400% composite graft is equivalent to a single layer 400% elongation graft and substantially stronger than a single layer 900% elongation graft, despite an overall thinner wall dimension. Additionally, the superior strength of the composite graft is demonstrated by the higher elongation capable of being borne by the graft prior to failure. The lower suture hole elongation, indicative of a smaller tear being caused by suturing and tensioning at a fixed value of 100 grams is clearly demonstrated for the graft prepared by the method of the current invention.

TABLE 1

Physical Property Measurement	400% Elongation Single Layer Graft	900%/400% Elongation Composite Graft	900% Elongation Single Layer Graft
Radial Tensile Strength (kg/mm ²)	0.48	0.48	0.2
Radial Strain at Break (%)	550	690	531
Suture Hole Elongation (%)	87	81	158
Wall Thickness	0.72	0.65	0.73

EXAMPLE 3

Three composite grafts were constructed and their ability to reseal after a puncture was tested. Graft No. 1 is an ePTFE helically tape-wrapped graft with no resealable layer, and graft Nos. two (2) and three (3) were constructed with a resealable intermediate layer in the below-described procedure.

Graft No. 1 is an ePTFE graft used as the control in the following experiment. Graft 2 was constructed by first placing an ePTFE tubular structure with an inner diameter of 5 millimeters on a mandrel. A thermoplastic elastomer tubing was then slid over the ePTFE tube.

Because of the tackiness of the thermoplastic elastomeric tubing, it was necessary to manipulate the tubing by rolling and stretching it in order to maneuver it over the first ePTFE tubular structure to lie evenly thereon. A second ePTFE tubular structure with an inner diameter of five (5) millimeters was then radially stretched, or expanded to yield an ePTFE tubular structure with an inner diameter of 8 millimeters. A metal sleeve was then placed over the thermoplastic elastomer-covered tubular structure. The second 8 mm inner diameter ePTFE tubular structure was then placed onto the metal sleeve. The metal sleeve was then retracted allowing the second tubular structure to come into contact with the thermoplastic elastomer.

Both grafts Nos. 2 and 3 were prepared using this procedure. Graft 2 was then placed in an oven and heated at 350°F for 10 minutes. There appeared to be little bonding between the layers, and the graft was then heated for an additional 10 minutes at 450°F.

Graft 3 was heated for five minutes at 400°F. The material did not appear to bond together, so the graft was then heated at 425°F for an additional five minutes.

Grafts Nos. 1-3 were then tested for quality control. Each graft was placed in a water entry pressure measuring device to provide a constant pressure of water within the graft. The water pressure was maintained at three pounds per square inch. The grafts were then punctured with a 20 gauge needle. The needle was then removed and any water leaving the puncture site was collected in a beaker and measured over the time of collection. This procedure was followed for a number of test runs. The results are shown in table two below.

TABLE 2

Run Number	Graft No. 1 (grams water/30 seconds)	Graft No. 2 (grams water/30 seconds)	Graft No. 3 (grams water/30 seconds)
1	21.7	0.2	1.2
2	20.0	1.3	0.5
3	23.1	0.2	—
4	20.0	1.4	—
5	21.9	—	—
Average Of Trial Runs	21.34	0.775	0.85

5 Graft No. 1 leaked steadily, as the puncture did not reseal. While graft numbers 2 and 3 showed minimal leakage. In both cases, the water leaked in a small trickle in the first few seconds (2-5 seconds), then stopped or slowed to an immeasurable seepage.

EXAMPLE 4

10 Three additional grafts were constructed with an intermediate resealable layer and tested to determine their ability to seal after puncture.

15 Graft No.4 was made using an ePTFE graft with an initial 5 millimeter inner diameter as the inner tubular structure. The 5 mm ePTFE tube was extruded using a die insert of 0.257 inches, and a mandrel of 0.226 inches. The inner tube was then stretched longitudinally to 500% its original length, and sintered. The inner tube was then stretched radially by placing it on a 5.95 mm mandrel. A pre-sintered PTFE bead-wrap (0.014 inch \pm 0.002 inch diameter, length approximately 30 cm) was then helically wrapped around the exterior of the inner tube at 650
20 revolutions per minute (RPM), and with a mandrel speed transversely at 800 RPMs. The wrap helically repeated on the tube at approximately every 3.5 cm, and at an angle of approximately

10°-50° with respect to a radial axis. After wrapping, the coil wrapped tube was then heated in an oven at 663°F for 10 minutes to sinter the beads to the grafts.

5 A second ePTFE tube was then added exterior the beaded coil and inner tube. The second ePTFE tube also had a 5 millimeter inner diameter, and was a tube extruded using a die insert of 0.271 inches and mandrel size 0.257 inches, and was stretched to longitudinally 500% its original length and sintered. The second ePTFE tube was then radially stretched over a tapered 6-10 mm mandrel to a graft of inner diameter of 10 mms. The second graft was then transferred to a 10 mm hollow mandrel within which the inner bead-wrapped graft was placed. The hollow mandrel
10 was then slid out to leave the second ePTFE tube exteriorly placed on the bead covered graft. The composite device was then placed in an oven for 20 minutes and heated at 663°F. Three 3 cm long tubular rings of thermoplastic elastomer (C-FLEX®) were then placed over the composite stent graft. The rings had an inner diameter of 6.5 mms. A third outer ePTFE tubular structure was then placed over the C-FLEX rings. The third outer tubular structure was a 3 mm
15 inner diameter ePTFE extruded tube was made using a die insert of 0.189 inches, and was stretched longitudinally to 700% its original length and sintered. The third outer structure was then expanded radially by stretching the tube over a 4-7 mm continuously tapered mandrel, then further radially expanded over a 6-10 mm continuously tapered mandrel. It was then placed on a 10 mm hollow mandrel with the C-FLEX covered composite graft placed within the mandrel, and
20 the outer tube then covers the C-FLEX® upon removal of the hollow mandrel. Graft 4 was then heated for 5 minutes at 435°F.

25 Graft No. 5 was constructed using a 5 mm inner diameter ePTFE extruded tube as the inner tubular structure. The 5 mm tube was then radially stretched over a 5.95 mm diameter mandrel. A corethane intermediate layer was then spun into a tube of 6 mm diameter with 250 passes, and subsequently loaded onto the inner ePTFE tube. A second outer ePTFE tubular structure (originally a 5 mm inner diameter) was then stretched to 10 mm over a 6-10 mm continuously tapered mandrel. Graft No. 5 was then heated for 15 minutes at 400°F.

Graft No. 6 was made using a 5 mm inner diameter ePTFE graft as the inner tubular structure. The inner tube was then stretched over a 6.2 mm hollow expansion mandrel. An ePTFE yarn was then wrapped in two helical directions over the inner tube. A corethane layer was then spun over the yarn wrapping as the third tubular layer at a mandrel speed of 1,000 RPMs and a wrap angle of 50 degrees. The corethane covered graft was then heated at 230°F for 10 minutes. An outer ePTFE tube with a 5 mm inner diameter was then stretched to 10 mm over a 6-10 mm continuously tapered mandrel. It was then mounted on the corethane layer with the use of the hollow mandrel. Graft 6 was then heated for 30 minutes at 340°F.

Grafts 4-6 were all tested by puncturing them with a constant pressure water source attached to them as done with Grafts Nos. 1-3. The results are shown in Table 3 below.

TABLE 3

Run No.	Graft 4 (grams water/seconds)	Graft 5 (grams water/seconds)	Graft 6 (grams water/seconds)
1	0.2	0	0
2	0	0	0
3	0	0.7	0

Various changes to the foregoing described and shown structures would now be evident to those skilled in the art. Accordingly, the particularly disclosed scope of the invention is set forth in the following claims.